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# Contribution of the support limb in control of angular momentum after tripping

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**Abstract**

Tripping over an obstacle can result in a fall when the forward angular momentum, obtained from impact with the obstacle, is not arrested. Angular momentum can be restrained by proper placement of the recovery limb, anteriorly of the body, but possibly also by a reaction in the contralateral support limb during push-off. The purpose of this study was to quantify the extent to which the support limb contributes to recovery after tripping by providing time and clearance for proper positioning of the recovery limb, and by restraining the angular momentum of the body during push-off. Twelve young adults were repeatedly tripped over an obstacle during mid-swing, while walking over a platform. Kinematics and ground reaction forces at the support limb were measured. Quantification of the angular momentum was based on calculation of the external moment, which equals the rate of change in the angular momentum of the body. Results showed that all subjects acquired a similar increase in angular momentum during foot–obstacle contact, on average $11.4\,\mathrm{kg\,m^2\,s^{-1}}$. In all subjects, the support limb played a role in recovery after tripping by providing time and clearance for proper positioning of the recovery limb, as indicated by body elevation (6%) and the increased forward pelvis displacement over recovery stride (43%). Almost all subjects were also able to restrain the forward angular momentum of the body during push-off by the support limb. Less angular momentum remained to be further accomplished by the recovery limb. Reductions in the quality of the support limb responses may be among the factors that increase the risk of falling in the elderly.


*Keywords:* Tripping; Falling; Angular momentum; External moment; Obstacle–foot contact force

## 1. Introduction

Falls and fall-related injuries cause serious problems for the growing population of the elderly. One in three adults over 65 years of age falls once a year, mostly as the result of a trip or slip (Nevitt et al., 1991; Berg et al., 1997; Ryynanen et al., 1991). The need to discover mechanisms underlying trip-related falls has led to several investigations of tripping (Grabiner et al., 1993; Eng et al., 1994; Grabiner et al., 1996; Schillings et al., 1999, 2000; Pavol et al., 2001; Smeesters et al., 2001).

The main purpose of the recovery reaction after tripping is to arrest the forward angular momentum, which the body gets from impact with the obstacle. An inadequate reaction will lead to a fall. Eng et al. (1994) described two phase-dependent modes of recovery reactions. Impact during early swing leads to an elevating strategy, in which the obstructed (ipsilateral) swing limb is lifted over the obstacle immediately after collision and placed forward, over the obstacle. Impact during late swing induces a lowering strategy, in which the obstructed foot is placed quickly before the obstacle and the other limb is subsequently placed anteriorly of the body. For both strategies, we call the limb that is placed anteriorly of the body the recovery limb, while the contralateral stance limb is called the support limb.

Placing the recovery limb anteriorly of the body is one means to reduce the angular momentum of the body (Grabiner et al., 1993, 1996; Pavol et al., 2001). This limb can generate a force and moment that counteract the angular momentum, provided that it is properly placed anteriorly of the body. Proper placement of the recovery limb can only be achieved if there is sufficient

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time and clearance. This can be brought about by rapid responses in the recovery limb itself, but in addition, the support limb can help to gain time and clearance by elevating the body during push-off.

In theory, the support limb can also contribute to recovery in another way, namely by reducing the forward angular momentum of the body during push-off, before the recovery limb hits the ground. Angular momentum can be controlled by generating adequate joint moments, and the associated rate of change in angular momentum is reflected in the external moment ($M_{ext}$), which is the moment of external forces about the body center of mass.

The purpose of this study was to determine whether the support limb contributes to recovery after tripping, and if so, to quantify the extent to which it contributes. We hypothesized the support limb to contribute in two ways: (a) by providing time and clearance for proper positioning of the recovery limb, and (b) by restraining or reducing the forward angular momentum of the body induced by the trip. The first role would be reflected in an increased upward and forward displacement of the pelvis during the push-off phase in tripping as compared to normal walking. The second role would be reflected in a sign change in the external moment during the push-off phase.

## 2. Methods

Twelve volunteers (6 male, 6 female) with a mean age of 27 years (SD 4) participated in this study. Subjects were informed on the research procedures before they gave informed consent in accordance with the ethical standards of the declaration of Helsinki. Participants walked approximately 60 times over a platform in which 21 obstacles were hidden. In about 10 trials, the subjects were tripped over one of these obstacles. A computer controlled, based on online kinematic data, which one of these obstacles had to appear at what time, so as to cause a trip at mid-swing, allowing us to focus on the elevating strategy. In addition to kinematics, we measured ground reaction forces of the support limb. Details on the experimental setup and protocol are described below.

Subjects, wearing walking shoes, were instructed to walk at a self-selected speed over a platform of 12 m. In the platform, a force plate was mounted and 21 aluminum obstacles of 15-cm height (28.5-cm width) were hidden over a total distance of 1.5 m (Fig. 1). In about 10 out of 60 walking trials, one of the obstacles suddenly appeared to trip the subject, either on the left- or the right-hand side. At the start of each trial, subjects did not know whether, or where an obstacle would appear. Online kinematic data of each trial were used to calculate the subject's step length and velocity. Based on these variables, position and timing of the obstacle to appear were chosen, so as to cause a trip at a certain percentage of the swing phase. Given the inter-obstacle distance of 7 cm, the obstacle appeared within 3.5 cm of the calculated position. The experimenter controlled whether or not an obstacle should appear, at which side (left or right) and at which percentage of the swing phase. In this experiment, at least 5 trips were evoked to trip the subject on the left limb at mid-swing to obtain comparable reactions (elevating strategy) and collect the ground reaction force data while the support limb was on the force platform. A full-body safety harness, attached to a ceiling-mounted rail, ensured that subjects would not be injured should their recovery reaction be inadequate. The safety ropes provided enough slack for free motion and harness assistance could be precluded visually, to which end all trials were recorded on video.

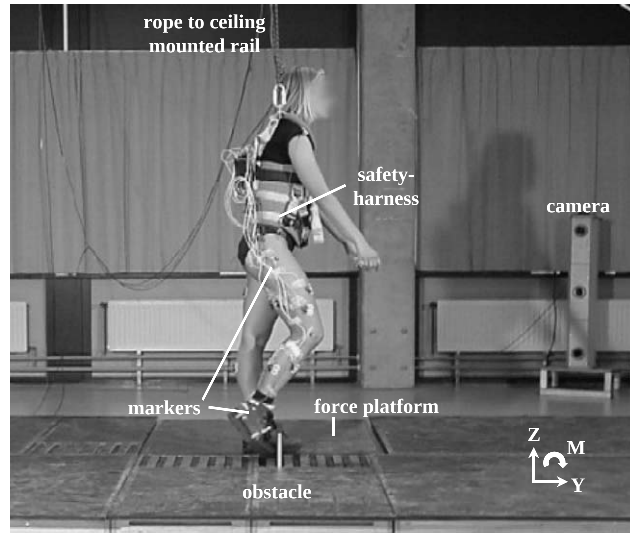


Fig. 1. Picture of the experimental setup. A force platform and Optotrak cameras were used for data collection of kinetics and kinematics. Twenty-one obstacles were hidden in the floor. One obstacle could suddenly appear, based on kinematic data of the ongoing trial, to trip the subjects at a specific time. Subjects wore a safety-harness.

Gait kinematics were recorded during each trial using 4 Optotrak cameras (Northern Digital ©). Motion of 12 infrared-light emitting markers was tracked. The markers were placed bilaterally over the anatomical landmarks heel, metatarsophalangeal joint (MTP5), lateral malleolus, lateral epicondyle and trochanter major of the femur, and acromial process. The coordinates of these landmarks defined 7 body segments: 2 feet, 2 lower legs, 2 upper legs and a head–arms–trunk (HAT) segment. Ground reaction forces at the right foot were recorded by a custom-made strain gauge force plate ($1 \times 1$ m). From the distribution of the force components, the center of pressure (COP) was calculated. LabVIEW (National Instruments ©) was used to synchronize and collect the kinematic data and ground reaction forces at a sample frequency of 100 Hz and to control the appearance of obstacles hidden in the walkway (see above and Oudejans and Coolen, 2003).

For each subject, 5 normal walking trials and 5 left leg tripping trials at mid-swing were randomly selected from successful trials with complete kinematic and dynamic data. In 2 subjects, complete data of only 3 tripping trials was available. Heel strike (HS) and toe-off (TO) were detected on the basis of kinematic data, as force plate data were not available for the left foot. HS coincided with a local minimum in the vertical velocity component of the toe marker and TO coincided with a local maximum in the vertical velocity component of the heel marker (Pijnappels et al., 2001). Impact (or contact) of the foot with the obstacle coincided with a local minimum in the acceleration of the toe marker in the walking direction. Based on HS, TO and obstacle–foot contact events, data were analyzed in the sagittal plane after smoothing with a one-directional second order low-pass Butterworth filter with a cutoff frequency of 8 Hz. One-directional filtering preserved the timing of the start of obstacle–foot contact onto the data.

To investigate the contribution of push-off by the support limb in gaining time and clearance for proper positioning of the recovery limb, we calculated body elevation (hip height) and timing parameters. Hip height was calculated as the height of the bilateral hip markers, relative to subjects' hip height at HS. For timing parameters, we calculated duration of stride (from HS until HS), stance phase (from HS until TO), swing phase (from TO until HS) and double support phase (from HS of the one limb until TO of the other limb). For statistical analysis of differences in these parameters between normal walking and tripping reactions, within-subject averaged (across trials) values were analyzed in a multivariate analysis of variance (MANOVA) for repeated measures. The level of significance was set at $p<0.05$.

The contribution of the support limb to restrain angular momentum of the body during push-off was investigated by calculating the external moment ($M_{\text{ext}}$), which equals the rate of change in the angular momentum of the entire system. Calculation of angular momentum directly from the kinematic data was not deemed to be very accurate, because the angular momentum of arm segments, which made vigorous flexion and endorotation, could not be determined. Therefore, $M_{\text{ext}}$ was used as a measure for the rate of change in angular momentum.

$M_{\text{ext}}$ was calculated as the sum of the moments generated by external forces acting on the system:

$$M_{\text{ext}} = \frac{\mathrm{d}\sum(I\omega)}{\mathrm{d}t} = \vec{F}_{\text{gr}} \times \vec{d}_{\text{gr}} + \vec{F}_{\text{c}} \times \vec{d}_{\text{c}},$$

where $\vec{F}_{\text{gr}}$ is the ground reaction force at the COP, $\vec{F}_{\text{c}}$ is the contact force of the obstacle at the toe, $\vec{d}_{\text{gr}}$ and $\vec{d}_{\text{c}}$ are the vectors from the body center of mass (COM) to the point of application of the respective force vectors (Fig. 2). GRF was measured directly by the force platform. The obstacle–foot contact force ($F_{\text{c}}$) was calculated from the linear impulse over a period from 10 ms prior to impact to return of the foot–obstacle contact force to 0 N. During this phase, the external contact force equals the difference between the rate of change in linear impulse and the GRF plus force of gravity:

$$\vec{F}_{\text{c}} = \frac{\mathrm{d}(m_{\text{body}} \cdot \dot{\vec{R}}_{\text{COM}})}{\mathrm{d}t} - \vec{F}_{\text{gr}} - m_{\text{body}} \cdot \vec{g},$$

where $\vec{F}_{\text{c}}$ and $\vec{F}_{\text{gr}}$ are the contact forces and GRF, respectively, $m_{\text{body}}$ is body mass, $\dot{\vec{R}}_{\text{COM}}$ is the linear velocity of the body COM and $\vec{g}$ is $-9.81$. The contact phase was followed by the push-off phase, which is defined as the period from the end of foot–obstacle contact to the end of the single support phase.

For determination of the vectors $\vec{d}_{\text{gr}}$ and $\vec{d}_{\text{c}}$, position of body COM was calculated from the segments' masses and center of mass locations. The inertial parameters of each segment (mass, position of the segmental center of mass and the segmental moment of inertia) were calculated per subject, according to Plagenhoef (Plagenhoef et al., 1983). The HAT was represented as a single link from the bilateral hip joint centers to the HAT COM. The position of the HAT COM was calculated by using the criterion that the reactive forces acting at the hips, calculated by inverse dynamics, equaled the force necessary for (translational) acceleration of the

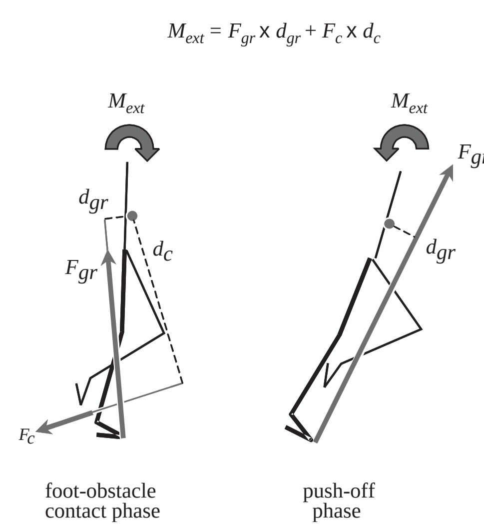


Fig. 2. External moment ($M_{\text{ext}}$) was calculated as the sum of moments of the external forces about the body center of mass (•). $F_{\text{gr}}$ is the ground reaction force and $F_{\text{c}}$ is the foot–obstacle contact force, $d_{\text{gr}}$ and $d_{\text{c}}$ (dashed lines) are the moment arms of the respective force vectors. During foot–obstacle contact phase, the moment effect of $F_{\text{gr}}$ and $F_{\text{c}}$ on the body indicates an increase in forward angular momentum, whereas during push-off, theoretically, a decrease in angular momentum can be achieved.

HAT segment:

$$\vec{\ddot{R}}_{\mathrm{COM}} = \frac{\vec{F}_{\mathrm{hips}} + m_{\mathrm{body}} \cdot \vec{g}}{m_{\mathrm{body}}},$$

where $\vec{F}_{\mathrm{hips}}$ is the force acting at the hips. This way, we calculated the acceleration of the HAT COM, resulting in zero residual forces. Velocity and position of HAT COM were calculated by integration. For initial conditions we used velocity and position of a HAT COM on the line between hip and shoulder joint centers at the first sample of the single support phase. Calculation of HAT COM was limited to the single support and aerial phases, as the external GRF was only available for this period.

## 3. Results

Tripping reactions were induced on average at 39 (SD 3.8)% of the normal swing phase duration. Typically after tripping in this particular phase of the gait cycle, subjects performed an elevating strategy. Fig. 3 depicts stick diagrams of two typical subjects (2 and 9) for both normal walking and tripping. Table 1 represents the general parameters for both normal walking and tripping. The duration of a stride, normally 1.03 (SD 0.04) s, was increased significantly for the obstructed swing (recovery) limb as well as for the support (push-off) limb. The increase in stride duration was attributed to an increase in stance phase duration of the push-off limb (13%), and to an increase in swing phase duration of the recovery limb (63%). The double support phase was not present after tripping. Instead, an aerial phase was seen. These findings indicate that extra time was available for positioning of the recovery limb. Furthermore, the stride length of the recovery limb was increased (10%). Stride length can be determined by actions of both the support limb and the recovery limb, but horizontal displacement of the pelvis over the recovery stride (i.e., from toe-off until landing of the

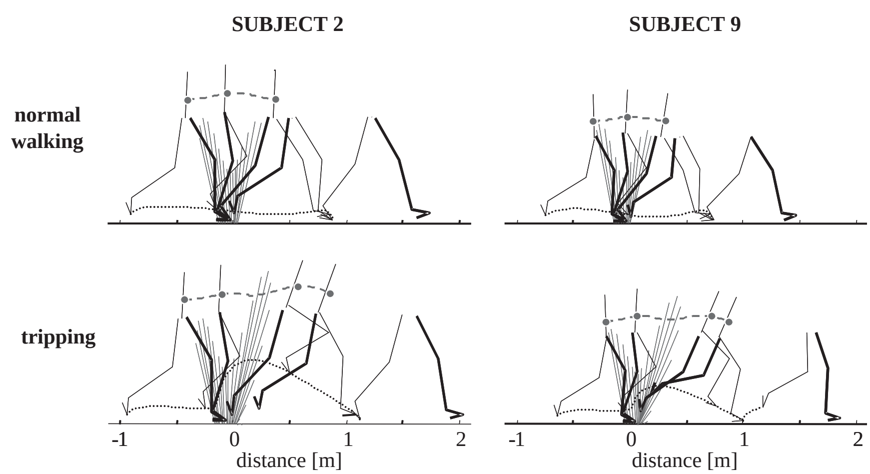


Fig. 3. Stick figures of two subjects (left or right columns) during a typical walking trial (upper graph) and a tripping trial (lower graph) for 5 instants of time (right and lift toe-off and heel strike, mid-swing or trip initiation). The obstructed (left) swing limb is indicated by thin lines, thick lines depict the contralateral support limb. The HAT segment is defined from the bilateral hip joint centers to the (optimized) location of HAT COM. Ground Reaction Force vectors, body COM position (•) and trajectory (dashed line), as well as toe trajectory over time (dotted line) are drawn. Note that HAT COM could not be calculated during double support.

Table 1
General parameters (timing, stride length and bilateral hip height) during normal walking (averaged over limbs) and tripping (for support limb and obstructed swing limb separately). Averages (and SD) over 5 trials and 12 subjects. Negative double support indicates an aerial phase. Statistical significance at $p<0.05$

| | Normal walking: left & right limbs | | Tripping: support limb (push-off) | | Tripping: swing limb (recovery) | | |
|---|---|---|---|---|---|---|---|
| Velocity (m/s) | 1.61 | (0.15) | 1.61 | (0.17) | 1.44 | (0.14) | a b c |
| Frequency (steps/min) | 117 | (4.50) | 109 | (12.27) | 96 | (7.71) | a b c |
| Cycle time (s) | 1.03 | (0.04) | 1.12 | (0.12) | 1.26 | (0.10) | a b c |
| Stance phase (s) | 0.61 | (0.03) | 0.69 | (0.07) | 0.61 | (0.03) | a b c |
| Swing phase (s) | 0.44 | (0.02) | 0.45 | (0.08) | 0.67 | (0.09) | a b c |
| Double support (s) | 0.09 | (0.01) | 0.09 | (0.01) | −0.05 | (0.05) | a b c |
| Stride length (m) | 1.66 | (0.15) | 1.81 | (0.26) | 1.83 | (0.24) | a |
| Hip height at toe-off (m) | 0.86 | (0.04) | | 0.91 | (0.04) | | a |
| Hip displacement over stride (m) | 0.83 | (0.08) | | 1.19 | (0.16) | | a |

[a]Significant difference between conditions; [b]Difference between sides; [c]Interaction condition × side.

recovery foot) can only be achieved by actions in the support limb. The bilateral hip displacement was increased by about 43% after tripping (Table 1). Furthermore, the body was elevated during push-off, as can be seen in Fig. 4. During normal walking, the position of the bilateral hip joint markers is highest in mid-stance and lowest in the double support phase, whereas after tripping, the body was elevated additionally during push-off by the support limb. This was seen in all subjects. At the end of push-off, the averaged hip height was about 5 cm higher after tripping compared to normal walking (Table 1). Typical obstacle–foot contact forces are presented in Fig. 5. Contact duration, averaged over trials and subjects, was 115 (SD 20) ms (Table 2). The horizontal (fore-aft) peak force was on average −177 (SD 43) N. The vertical force showed in all subjects a maximum of on average 48 (SD 24) N, followed by a minimum of −84 (SD 45) N.

During normal walking, a propelling $M_{\mathrm{ext}}$ is generated during push-off and an upright position is maintained by a counteracting force at heel strike of the next step. Successive positive and negative excursions in $M_{\mathrm{ext}}$ cancel each other over a stride cycle. Until trip initiation, there was of course no difference in $M_{\mathrm{ext}}$ between the walking and tripping conditions. During obstacle–foot contact, the body started rotating forward (clockwise), due to the external contact forces and gravity. The increase in angular momentum is reflected in a positive $M_{\mathrm{ext}}$ (i.e., angular acceleration of forward rotation). In all subjects, the area under the curve (AUC) of $M_{\mathrm{ext}}$, which equals the angular momentum, was increased over obstacle–foot contact phase, by on average 11.4 kg m$^2$ s$^{-1}$ (Table 2). Angular momentum can be controlled by generating adequate joint moments during push-off; the decrease in angular momentum would be reflected in a negative external moment. When considering the contribution of the support limb to recovery of the angular momentum, substantial between subject variations were noted, although the reproducibility within-subjects seemed very high (see SD in Fig. 6 and Table 2). Three subgroups could be defined, based on the capacity to restrain the angular momentum during push-off (Table 2). For the two major subgroups, individual data of two representative subjects will be presented first and described in detail. Fig. 6 presents the external moments and the integral of $M_{\mathrm{ext}}$, averaged over trials of these two exemplary subjects. $M_{\mathrm{ext}}$ in subject 2 became negative during push-off, indicating that the angular momentum is reduced. In this subject, the forward rotation that the body acquired during impact is completely eliminated during push-off (Table 2). Subject 9 stopped the increase in angular momentum as well, but did not manage to reduce it. This subject needed an extra step of the recovery limb to fully eliminate the forward rotation of the body. Indeed, Fig. 3 shows another jump and aerial phase in the following step for subject 9 whereas subject 2 had regained a normal walking pattern in the subsequent step. The outcomes of the reactions by the two subjects presented were representative for the main subgroups (Table 2 and Fig. 7). Elimination of the angular momentum was achieved by 4 subjects (numbers 1–4) and reduction was achieved by 6 subjects (numbers 5–10). Two subjects (11 and 12) did not react adequately during push-off; their angular momentum continued to increase over the whole push-off duration. Still, none of the subjects fell

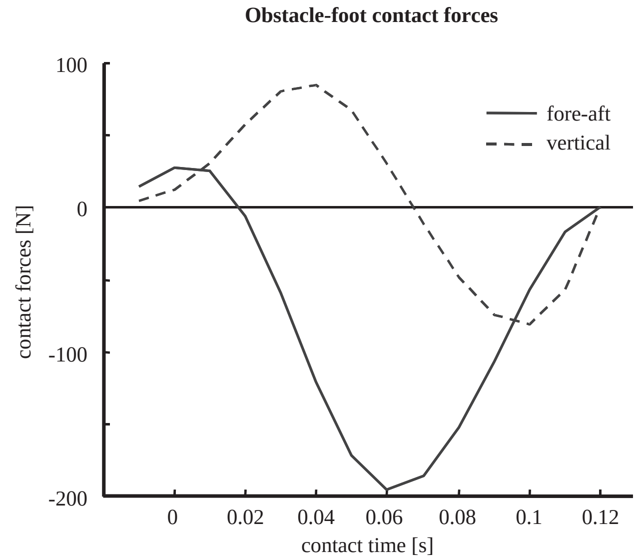


Fig. 5. Obstacle–foot contact forces (fore-aft and vertical) from 10 ms prior to impact to return of contact forces back to 0 N, for one typical tripping trial.

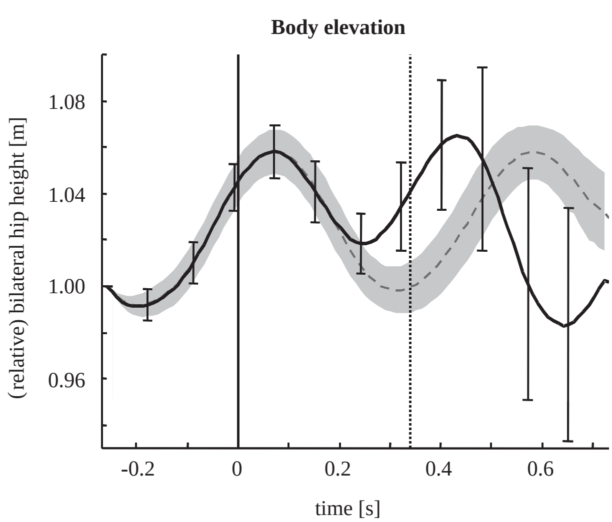


Fig. 4. The relative height of the bilateral hip markers as indication for body elevation during push-off, averaged over trials and subjects, and relative to subjects' hip height. Mean graphs over complete stride (from heel strike to heel strike) for normal walking (dashed mean and shaded SD) and for tripping (solid line and SD in error bars). Vertical lines indicate trip initiation (solid) and end of single support phase (dotted).

Table 2
Duration (ms) of obstacle–foot contact (from collision to obstacle free), push-off (from obstacle-free to end single support phase) and the sum of both phases. Averages (and SD) per subject over 5 trials. Area under the curve (AUC, $kg\,m^2\,s^{-1}$) of $M_{ext}$ over obstacle–foot contact phase, push-off phase and the sum of both phases. Subject 1–4 were able to fully reduce the increased angular momentum (negative, counterclockwise AUC of $M_{ext}$ during push-off), subject 5–10 restrained the increase (AUC of about $0\,kg\,m^2\,s^{-1}$), and subject 11 and 12 were not able to restrain during push-off (further positive AUC of $M_{ext}$ over push-off)

| Subject | Contact duration | | Push-off duration | | Total duration | | Contact AUC $M_{ext}$ | | Push-off AUC $M_{ext}$ | | Total AUC $M_{ext}$ | |
|---|---|---|---|---|---|---|---|---|---|---|---|---|
| 1 | 96 | (23) | 342 | (22) | 438 | (12) | 11.2 | (2.4) | −14.5 | (1.4) | −3.3 | (2.2) |
| 2 | 120 | (26) | 254 | (20) | 374 | (21) | 14.1 | (3.8) | −12.8 | (3.8) | 1.3 | (1.6) |
| 3 | 108 | (15) | 194 | (44) | 302 | (36) | 14.0 | (2.4) | −7.8 | (3.5) | 6.2 | (5.7) |
| 4 | 120 | (13) | 228 | (41) | 348 | (33) | 8.5 | (1.6) | −6.3 | (4.6) | 2.2 | (3.1) |
| 5 | 110 | (15) | 262 | (30) | 372 | (15) | 12.1 | (1.8) | −1.9 | (4.3) | 10.2 | (5.8) |
| 6 | 114 | (5) | 358 | (44) | 472 | (42) | 9.4 | (2.3) | −1.6 | (1.7) | 7.8 | (3.3) |
| 7 | 110 | (0) | 327 | (29) | 437 | (29) | 10.0 | (0.5) | 0.6 | (3.6) | 10.6 | (3.1) |
| 8 | 132 | (7) | 322 | (23) | 454 | (26) | 14.5 | (1.8) | 0.6 | (2.3) | 15.1 | (3.1) |
| 9 | 132 | (16) | 274 | (24) | 406 | (19) | 11.4 | (3.2) | 1.1 | (1.3) | 12.5 | (3.2) |
| 10 | 132 | (4) | 394 | (47) | 526 | (48) | 17.1 | (1.4) | 6.1 | (2.9) | 23.3 | (3.0) |
| 11 | 93 | (17) | 193 | (29) | 287 | (38) | 6.0 | (1.7) | 6.3 | (2.3) | 12.3 | (3.4) |
| 12 | 100 | (14) | 278 | (32) | 378 | (45) | 8.5 | (1.3) | 11.0 | (0.8) | 19.5 | (1.1) |
| All | 115 | (20) | 287 | (69) | 402 | (73) | 11.4 | (1.6) | −2.0 | (2.7) | 9.8 | (3.2) |

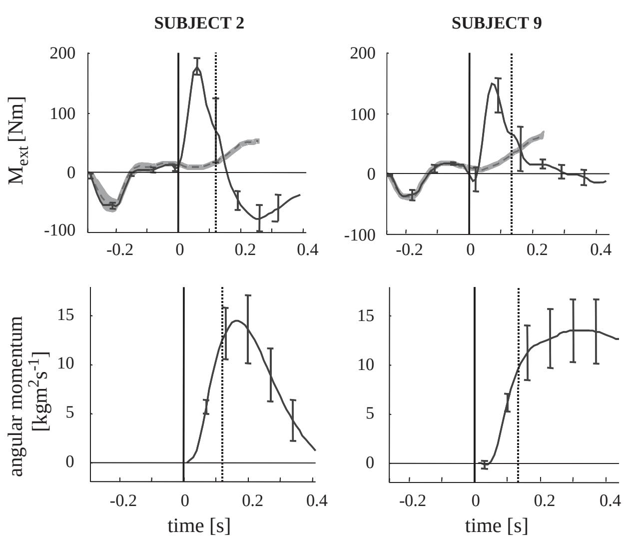


Fig. 6. $M_{ext}$ and integral of $M_{ext}$, which equals the angular momentum. Graphs from heel strike of the support limb until end of single support phase and averaged over 5 trials for normal walking (dashed mean and shaded SD) and for tripping (with error bars). Vertical lines indicate trip initiation (solid) and end of foot–obstacle contact phase (dashed). A positive $M_{ext}$ reflects an increase of angular momentum (clockwise acceleration), a negative $M_{ext}$ indicates a decrease of angular momentum (counterclockwise).

into the harness, so they were all able to recover eventually, although contribution of support limb was different and subsequent recovery steps were necessary in some subjects.

## 4. Discussion

This study revealed the contributions of the support limb to recovery after tripping. In all subjects, push-off generated by the support limb provided extra time and clearance for proper positioning of the recovery limb. In most subjects, the support limb additionally contributed by restraining or reducing the angular momentum of the body during the push-off. Up to date the literature on tripping has mainly focused on the swing limb (Eng et al., 1994; Grabiner et al., 1996; Schillings et al., 1999, 2000; Pavol et al., 2001). The present results suggest that support limb responses are functionally important and merit further investigation. Before discussing the role of the support limb in recovery after tripping, we need to address some methodological points.

The results presented here were based on experiments in which subjects were aware of the fact that they would be tripped in some trials. We have previously shown that this does not greatly affect gait kinematics (Pijnappels et al., 2001). The high reproducibility of the characteristics of the tripping responses in the present study (see Fig. 6 and Table 2) supports the idea the valid experimentation with respect to tripping responses is possible.

Estimation of the external moment ($M_{ext}$), which was used to study angular momentum, required knowledge the location of the body COM. As in earlier studies (Kingma et al., 1995) we used optimization methods to improve the position of the trunk COM to get a better body COM. Acceleration of the HAT COM was based on the reactive forces acting at the hips, which reflect (translational) accelerations of the HAT segment. In the first phase of single support, we assume no effect of arm swing on HAT COM and therefore we felt safe to use velocity and position of a HAT COM on a fixed point on the line between hip and shoulder joint centers (according to Plagenhoef et al., 1983) for initial conditions for integration.

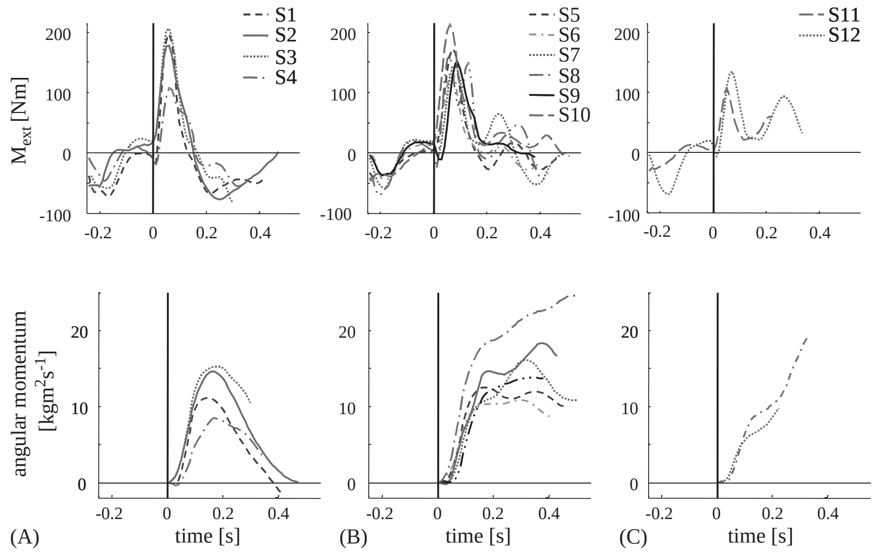


Fig. 7. $M_{ext}$ and angular momentum for all subjects, divided in the 3 subgroups (see table 2). (A) group of 4 subjects who were able to to fully reduce the angular momentum (angular momentum back to zero over push-off), (B) group of 6 subjects who restrained the increase in angular momentum (integral becomes constant), and (C) 2 subjects who were not able to restrain during push-off (further increase in angular momentum over push-off).

Another requirement for the validity of calculation of $M_{ext}$ was the determination of the obstacle–foot contact force. We based the calculation of these contact forces on the linear impulse. Over the duration of obstacle–foot contact, we expected no effects yet from arm movements on linear velocity of body COM. Recently, Zhou et al. (2002) measured obstacle–foot contact forces during walking, using a 3D-force platform. They found a contact duration of 90 ms, with a fore-aft and vertical maximum value of 129 and 49 N, respectively. Our calculations yielded similar contact duration and peak values of forces. Any difference might be caused by a difference in walking velocity and time of impact during the swing phase of this single trial measurement of Zhou et al. (2002).

Quantification of the angular momentum by calculation of $M_{ext}$ enabled us to investigate the contribution of the support limb to recovery after tripping. All subjects showed a similar increase in angular momentum during foot–obstacle contact. Provided proper (forward) positioning of the recovery limb, this limb can generate a force and moment that counteract the angular momentum of the body. After being tripped, all subjects showed an increase in stance duration of the support limb and swing duration of the recovery limb, an aerial phase instead of double support, as well as body elevation during push-off and elongation of the stride. Eng et al. (1994) also mentioned body height elevation during the elevating strategy. Body elevation started early in the push-off phase (Fig. 4). Rapid body elevation and forward propelling of the pelvis, together with the duration of stance, swing and aerial phase indicated that push-off by the support limb contributed to gaining time and clearance for proper placement of the recovery limb.

During push-off adequate joint moments in the support limb (reflected in $M_{ext}$) can also contribute to a reduction of angular momentum. Although results were very reproducible within subjects, different reactions were noted among subjects. Almost all subjects restrained the angular momentum after tripping by a reaction of the support limb, but not all subjects were able to actually reduce the angular momentum during this phase. The question remains what caused these differences between subjects. It could be due to initial conditions, such as walking velocity, trunk angle (and velocity) at time of tripping or joint moment generating capacity. However, no such differences between subjects in walking velocity or trunk angle were obvious. It seems, therefore, that the quality of the reaction in the support limb differs among subjects. Still, the present study showed that all subjects reacted very rapidly in an attempt to control the angular momentum. Further research on response times and response mechanics is required to investigate how an adequate push-off reaction is achieved.

The results of the present study show that the support limb plays an important role in recovery after tripping during push-off. For proper placement, the obstructed swing limb, of course, has to be swung forward. Mechanical requirements in the recovery limb, however, are first expected to become critical after landing when forces and moments have to be generated for counteraction of the angular momentum of the body. The support limb can provide enough time and clearance for proper positioning of the recovery limb. Furthermore, the more reduction in angular momentum achieved by the support limb during push-off, the less remains to be accomplished by the recovery limb. All subjects provided time and clearance during push-off. Most subjects were also able to restrain angular momentum of the body during the push-off by the support limb, some of them even completely reduced the forward angular momentum. Reductions in the quality of the support limb responses may be among the factors that increase the risk of falling in the elderly. Further research is needed to characterize these responses in both young and elderly subjects.

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